Stapp Car Crash Journal, Vol. 56 (October 2012), pp. 469-484

2012-13

An Integrated Helmet and Neck Support (iHANS) for Racing Car Drivers: A Biomechanical Feasibility Study

James A. Newman
Newman Biomechanical Engineering Consulting Inc., Edmonton, AB, Canada

Christopher Withnall and Michael Wonnacott
Biokinetics and Associates Ltd., Ottawa, ON, Canada

ABSTRACT – A new form of head and neck protection for racing car drivers is examined. The concept is one whereby the helmet portion of the system is attached, by way of a quick release clamp, to a collar-like platform which is supported on the driver's shoulders. The collar, which encircles the back and sides of the driver's neck, is held in place by way of the on-board restraint belts. The interior of the helmet portion of the assembly is large enough to provide adequate volitional head motion. The overall objective of the design is to remove the helmet from the wearer's head and thereby to mitigate the deleterious features of helmet wearing such as neck fatigue, poor ventilation and aerodynamic buffeting. Just as importantly, by transferring the weight of the helmet and all attendant reaction forces associated with inertial and impact loads to the shoulder complex (instead of to the neck), reduced head and neck injury probability should be achievable.

This paper describes the concept development and the evolution of various prototype designs. Prototypes have been evaluated on track and sled tested in accordance with contemporary head neck restraint systems practice. Also discussed is a series of direct impact tests. In addition, low mass high velocity ballistic tests have been conducted and are reviewed herein.

It is concluded that this new concept indeed does address most of the drawbacks of the customary helmet and that it generally can reduce the probability of head and neck injury.

KEYWORDS – racing, helmet, injury, iHANS, HANS, impact, protection, head, neck, feasibility

BACKGROUND

The use of helmets by auto race drivers has its origins in the days when restraint systems were nonexistent or ineffective and ejection from the vehicle in a crash was not uncommon. Even if not ejected, roll cage structures were not typically used and the driver's head could readily make contact with the roadway during a rollover. The driver's head in early racing car crashes was exposed to impact much the same as that of an ejected motorcyclist and protective headgear developed for motorcyclists was typically used. Figure 1 shows the degree to which the driver's head was typically exposed.

Figure 1: Sir Stirling Moss at the 1959 Grand Prix of Monaco wearing a classic "pudding basin". Photo from the Cahier Archive

Address correspondence to James A Newman, Newman Biomechanical Engineering Consulting Inc., 1152 119th Street, Edmonton, Alberta, Canada, T6J 7H6.
newman@biomechanical-engineering.com

Notwithstanding fairly recent developments in auto racing headgear (FIA 8860-2004), auto racing driver helmets are to this day, essentially motorcycling helmets.

The modern *crash helmet* typically comprises a stiff, strong fiber composite shell, an interior crushable liner (usually molded expanded polystyrene bead EPSB foam) and a soft pliable *comfort* liner. The shell serves to distribute the impact over a broad surface and resists penetration. The EPSB liner deforms and crushes to absorb impact energy and the comfort padding serves to insure that the helmet fits snugly to the wearer's head.

Due to improvements in roll cage, restraint system and cockpit design, direct impact with objects outside the car is infrequent. Head impact within the car is of course still possible particularly in crashes involving a substantial rear or lateral component. In such crashes, the cockpit surround or seat head restraint design will largely dominate head impact response. In frontal collisions, the face/forehead, if unrestrained, could strike the steering wheel assembly.

Since the mid 1990's, the driver's head in many open cockpit cars is firmly ensconced within the cockpit such as shown in Figure 2.

Figure 2: Mark Blundell at a 2000 CART Grand Prix in Nazareth Pennsylvania. Michael Levitt photo courtesy of LAT photography.

In light of newer seat designs having energy absorbing structure surrounding and supporting the driver's head (Melvin and Gideon, 2004), the exact role of the helmet in dealing with closed head injuries in this environment is unclear (Grohs and Archer, 2000).

In any crash scenario, the driver's head can be struck by flying debris including car parts such as wheels, suspension components, etc.

Henry Surtees, the son of 1964 Formula 1 world champion John Surtees, died after being struck in the head during a Formula 2 race at Brands Hatch (in Kent, England) on 20 July 2009. The 18-year-old was knocked unconscious after being struck by a tire from another car. His more serious head injuries occurred when he subsequently crashed into a retaining barrier.

HEAD AND NECK INJURIES

In most cases, concussion, and other forms of closed head injury, are due principally to deceleration of the vehicle as a whole and movement of the driver within the vehicle structure; the latter being controlled by the coupling between the driver and the vehicle via the cockpit design and the restraint system (Melvin, 1998). It is generally acknowledged that closed head injury is due to head motion and that an understanding of the nature, severity and distribution of such motion must consider both linear and angular kinematics (King et al, 2003 for example).

Though injuries due to impact with projectiles on the race circuit are rare, such an impact can lead to localized loading and subsequent deformation or even structural collapse of the helmet shell. This results in concentrated loading on the wearer's head and the possibility of skull fracture and accompanying focal brain injuries. In 2009 during qualifying for the Hungarian Gran Prix, Felipe Massa was struck on the head by a spring that had come from the car ahead driven by Rubens Barrichello. He sustained life threatening head injuries but has since recovered and continues to compete.

Another uncommon but serious injury occurs to the neck or base of the skull when tensile or compressive loading on the neck becomes excessive.

The fatal accident of NASCAR driver Dale Earnhardt in February 2001, led to the current widespread adoption of a device intended to limit excessive head motion during a crash – the HANS® device. It couples the helmet of the wearer to the vehicle via a set of supplementary restraints straps attached to the helmet and the on board seat belt system. See Figure 3.

Figure 3: Jenson Button dons his helmet with the attached HANS system prior to the 2006 Grand Prix of Turkey. Photo from the Cahier Archive.

By controlling head flail the system reduces tensile neck strain during a crash. In addition, it provides the opportunity to reduce head excursion and thereby may reduce the likelihood of head impact and closed head injury.

HANS® was developed many years before Earnhardt's accident. Hubbard (1987) recognized the need to better control head motion principally to reduce the likelihood of high cervical injury. That it could also limit head excursion thereby preventing head/face impact was almost an added bonus. However, HANS®, or any contemporary head restraint system HRS, does little to reduce neck injury that may be associated with axial compression of the cervical column due to impact to the top of the helmet by, for example, an errant wheel. In addition, an HRS does little to reduce the demands on the wearer's head/neck associated with the normal wearing of the helmet itself. This is an issue of importance in the safe and effective operation of the car before any crash occurs. Drivers continue to tolerate neck fatigue, poor ventilation, discomfort, hearing issues, peripheral vision limits, etc., simply because of the custom of wearing a helmet. Coupling it with an HRS may make it work better in some crashes, but it doesn't make it inherently more effective or wearable.

CONCEPT DEVELOPMENT

Conceptually, the device comprises a helmet-like shell affixed to a shoulder/torso restraint assembly in such a way that the helmet mass, and the effects associated therewith, are borne by the driver's shoulders not his/her neck. Unlike the closely fitting contemporary helmet, the helmet portion of this design fits *over* the wearer's head leaving a *gap* around the head to permit relative movement of the head. Ideally, the wearer would have the same volitional head motion as he currently does when driving with a contemporary helmet/HRS. The *stand-off* between the head and the shell interior would greatly assist ventilation as well as voluntary head movement. The helmet portion may be readily detached from the collar via a quick release mechanism. The connection between the head, the shell and the shoulder assembly is such as to limit head motion under extreme loading conditions.

The overall attributes of this concept can be summarized as follows:

<u>Operational</u>

1. No helmet mass is borne by driver's neck thus less fatigue and more rapid head movement is possible.
2. Improved ventilation capability.
3. Potentially better streamlining relative to vehicle in open cockpit racing.
4. No aerodynamic buffeting of head at speed.
5. Easily donned compared to current helmet design.
6. Easily doffed by driver or crew.
7. Easy to provide drinking fluids to driver.
8. Eye glasses may be readily worn by driver.
9. Adequate driver head motion permitted.

<u>Emergency</u>

1. Easy rapid access to airway due to simple helmet removal procedure.
2. Helmet is readily removed without imposing any axial or bending loads on neck.
3. Driver extrication is not hampered in the way that a full canopy (as has been proposed) may.

<u>Crash Performance Targets</u>

1. Exceed relevant FIA helmet impact and penetration requirements.
2. Minimize compressive loading on neck when vehicle inverted or when object strikes head.

3. Provide comparable frontal crash injury reduction without the use of auxiliary straps and associated hardware attached to helmet.
4. Limit the extent of lateral motion of the head and neck in side impacts involving vehicles not equipped with head surround cushioning or extended seat backs.

CONCEPT DESIGN

Following study of driver's head motion requirements in an open wheel racing cockpit mockup, the required interior volume for head motion was established.[1]

Consideration for field of view and stand-off as well as room for some energy attenuating foam interior led to a required external geometry. These requirements were transformed via rapid prototyping into the first wearable mockup which is shown in Figure 4.

Figure 4: Initial concept mockup.

The *helmet* is supported by and attached to a modified HANS® collar. The in-car restraint belts are deployed over the shoulder flanges as is done with HANS®. This mockup was employed in a static environment to confirm head motion needs. (See Figure 5.)

[1] It is often the case that drivers can manage with much less head motion than that envisaged here. In oval courses or in drag racing for example, the driver's helmeted head is virtually fixed in place by the cockpit surround.

Figure 5: Checking head motion in mockup.

Though the overall appearance is similar to a contemporary helmet, it is considerably larger in this configuration. The eye-port is also very much wider than current designs. It was felt that these issues would be addressed in due course but it was not possible to assess driver reaction on-track since this is not a wearable (track-worthy) prototype. There was also concern that the appearance may be too *unusual*.

PROTOTYPE DESIGN

To address the above concerns, a new prototype was designed and assembled using an extra-large auto racing helmet as shown in Figure 6. With the interior comfort foam removed, this unit allowed a reasonable amount of head motion within the helmet. This particular prototype was designed to accommodate a reclined seating position for a Formula Ford driver with a medium size head.

Figure 6: First on-track prototype. (Elementary 1/4-turn clamps to facilitate attachment and removal of the upper section (helmet) were added.)

TRACK TRIALS

The first on-track tests were run in the summer of 2008 at Calabogie Motorsports Park in Ontario, Canada. The trials were recorded with on-board video and audio. (See Figure 7)

Figure 7: On track trials. (In this photo, the driver can be seen looking to his left within the helmet.)

The test driver's response was very positive. Several months later, a professional race car driving instructor wore another prototype in a closed Ford GT at Sebring International Raceway in Florida and his reaction was also very encouraging. Subsequent to these trials, one of the principals of HANS Performance Products Inc., Jim Downing, an experienced professional race car driver, ran with a prototype during practice runs at Roebling Road in Georgia.

The observations from these trials are summarized below:

1. Since no helmet mass is borne by driver's head or neck, more rapid head movement is possible. However, interaction between head and helmet interior can be distracting. When tilting head, especially when cornering on rough road, the driver's head bangs against the helmet interior.
2. It is apparent that good ventilation capability is possible. No wind noise related issues had arisen.
3. The elimination of aerodynamic buffeting of the head at speed in open cockpit cars reduces head/helmet vibration allowing a much sharper/stable visual field. There is also an absence of helmet lift.
4. The exaggerated eyeport of the first prototype is not necessary. Drivers have more than sufficient field of view with the customary helmet shape eyeport.
5. Because of the loose fit, it is easily donned compared to current helmet design. However the initial clamping system on the prototypes did not facilitate rapid self-donning/doffing of the unit.
6. There is a driver perception that the system may not be as safe since the head will crash into the helmet interior in an impact. This same concern has been voiced by investigators at NASA who are developing new suits and helmets for the astronauts of Orion Crew Module tol be used as part of the (since cancelled) Constellation Project. (Gohmert, 2009).
7. There is concern about capturing the chin/face in a frontal impact situation. (With HANS®, all restraining loads pass, in principle, through the forehead only). Drivers wondered how loading the forehead and the lower face might influence head/face injury patterns.

Notwithstanding the concerns expressed above, the concept was generally liked by all test drivers and the authors were encouraged to continue with the development and to establish the *safety* of the concept.

PROTOTYPE TESTING

The first objective of the tests was to determine if the prototype design could meet the current performance standard for head and neck restraints. In addition to assessing the experimental system's ability to moderate neck injury risk, tests were devised to assess the system's ability to mitigate head injury. Of particular interest was the influence of the gap between the head and the helmet interior on the helmet's ability to both control head motion and to attenuate impact between the head and the helmet interior.

To evaluate the protective nature of the experimental design, three different test methods were employed: Sled testing, low velocity high energy impact, and high velocity low mass ballistic impact. All three protocols employed some elements of the 50th percentile male Hybrid3 ATD.

The Hybrid3 was first introduced in 1976 (Foster et al 1977). Later, as newer restraint technology and injury assessment functions were being introduced, it became clear that the development of a more advanced frontal crash test dummy was appropriate (as reviewed in Haffner et al 2001).

Studies initiated by Melvin in (1988) led to the establishment of a whole set of requirements that were not being adequately addressed by the Hybrid3. This work led to the eventual development of the THOR (**T**est Device for **H**uman **O**ccupant **R**estraint) which was introduced at the 15th ESV Conference in 1996. The main features of THOR that the Hybrid3 does not possess that could have a bearing on the results of the present tests are:

- A thoracic cage that is more anthropometrically correct and which has more humanlike stiffness/damping characteristics.
- An articulating thoracic spine.
- A load bearing clavicular structure.

The above features can be important in the assessment of thoracic injury potential but, apart from whole ATD kinematic response, they are likely less important in the present tests. The relatively compliant nature of the shoulder structure of the Hybrid3 may, in the absence of load bearing clavicles, cause unrealistic reaction to the belt loaded collar which would otherwise be supported, at least partially, by clavicles. Furthermore the overly stiff somewhat under-damped Hybrid3 chest structure will cause unrealistic ATD rebound characteristics during sled tests.

Another biomechanical concern with the Hybrid3 is that associated with the design of the ATD neck structure. In the sled tests the only injury assessment being conducted is that of the neck, thus the biofidelity of this component has come under some scrutiny. In particular, the metal cable that runs through the structure may affect the accuracy of neck loading as well as the kinematic response in severe frontal crash testing. Investigators at Duke University have recently been retained to assist FIA to develop a more biofidelic neck to be used in evaluating head and neck restraint devices (Gramling, 2011).

In assessing injury potential, a variety of conventional and newer injury assessment functions are available. Those employed herein are listed along with supporting background in Appendix A. Though “injury risk curves” have been developed for many of these functions (Mertz et al 1996, for example) it is understood that all such functions provide no more than estimates of injury probability and as such their use in absolute terms is not advocated (except as they may be employed in performance standards). Thus in the present study, data is presented only in a comparative nature. That is, the response of the experimental system is simply compared to a standard contemporary helmet/HRS using the various numerical injury indices.

Sled Testing

Sled testing was conducted in accordance with the SFI 38.1 test protocol (2011). This is the test that all HRSs must pass in order to be considered in sanctioned racing. The protocol entails the use of a restrained Hybrid3 ATD subject to a velocity change of 63km/h (39.1mph) with a peak acceleration of 68G. The SFI 38.1 failure criteria for head and neck restraint systems are based solely on neck loading and are as follows.

Up to 80ms, neck compression and tension must be less than 2.5kN and the neck injury function Nij less than 1.0. Beyond 80ms, neck compression and tension may be as high as 3.2kN while Nij must remain less than 1.0.

The *late-in-the-pulse* criteria are more liberal and is a reflection of the “violent and unrealistic rebound of the (Hybrid3) ATD and the nature of the rigid seat structure used in the test.”(Melvin 2012). The crash pulse itself lasts about 60ms.

In the these tests, the Hybrid3 was seated in the specified “NASCAR” seat pan and restrained with a six point racing harness per SFI 38.1 specifications. In addition to the usual six-axes neck load cell instrumentation, the ATD head employed a tri-axial accelerometer at the head c of g. The headform also had a tri-axial load cell to monitor for mandible loading (See Figure 8). [2] Tests were run “head on” and at a 30 degree offset. In addition, the tests were recorded by high speed video.

[2] This instrument was developed by Biokinetics in conjunction with its work on mandible loading for the National Football League. This headform was assigned drawing number B-6070-D by Robert A. Denton, Inc. (later Denton ATD, now Humanetics).

Figure 8: Hybrid III headform with force-sensing mandible section.

Differing from the track version, this prototype (shown in Figure 9) was built to suit the more upright sitting position and employed a similar helmet shell and expanded polystyrene bead liner (sized for the Hybrid III head). Since there was no need to don and doff the helmet, it was bolted directly to the collar unit as shown. A circular viewing port was cut in the side of the helmet in order to observe the motion of the ATD head within. The width of the collar was increased so that the entire unit could be placed over the dummy head. In addition, a thick layer of expanded polypropylene bead foam was introduced in the chin bar area to assist capturing the lower face.

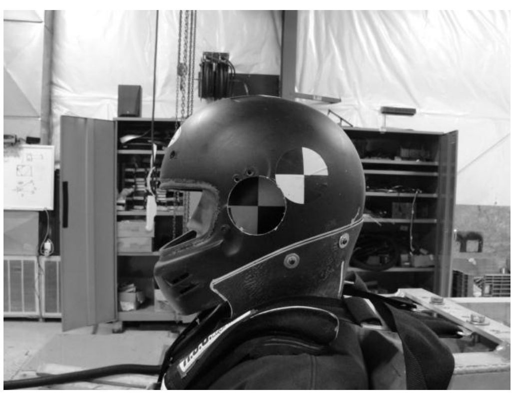

Figure 9: Sled test prototype.

These tests were among the first SFI 38.1 tests to be run at MGA Research Corporation and the sled pulse characteristics exceeded what is specified in SFI 38.1 having a peak acceleration of 69G and a velocity change of over 44mph.Though the prototype did not meet all the SFI 38.1 requirements in these tests, there were some interesting observations.

- Resultant head acceleration in both tests was less than 100G indicating good coupling to the sled and suggesting minimal risk of serious head injury.
- Maximum resultant loading to the mandible was approximately 500N. This is well below any expected lower facial injury tolerance level (Viano et al, 2004).
- In both the frontal and offset tests, the neck compression exceeded 2.5kN by a substantial margin. It was observed in the high-speed videos that compressive loading of the dummy neck begins as the belts snug down on the dummy shoulders pulling the unit down onto the crown of the head. Though the shoulder pads under the collar provide some compliance, thereby taking up some of the gap at the top of the head, it appeared that the relatively compliant character of the dummy shoulder structure was a significant contributor.

Based on these insights, a new test prototype was constructed and a second series of tests was run. Modifications to the prototype included stiffer padding between the collar and the ATD shoulders and relieving the liner thickness in the crown to provide increased space at the top of the ATD head.

In addition to the usual neck instrumentation, angular rate sensors were installed in the headform.[3] Linear head acceleration and mandible loading were not monitored this time. As before, a pure frontal and an offset 30 degree test were run. This time, the sled parameters were within the SFI 38.1 specifications. Comparative test results of the standard HANS® and iHANS are shown in Table 1 before 80ms and in Table 2 after 80ms.[4]

Table 1: SFI 38.1 responses before 80ms.

	frontal		30° offset		critical threshold
	Helmet w/HANS®	iHANS	Helmet w/HANS®	iHANS	
F_{zt} (kN)	0.2	0.5	1.8	0.5	2.5
F_{zc} (kN)	1.8	1.8	2.2	1.1	2.5
Max. N_{ij}	0.6	0.3	0.7	0.4	1.0

[3] Head acceleration is not recorded in SFI 38.1 testing. The test facility was asked to monitor angular head acceleration, and the only instrumentation available was angular rate sensors. However, the rotational data from the angular rate sensors proved to be unreliable and was not used.

[4] HANS® data was provided to the authors by HANS Performance Products Inc. and permission to publish the test results was given.

Table 2: SFI 38.1 responses after 80ms.

	frontal		30° offset		critical threshold
	Helmet w/HANS®	iHANS	Helmet w/HANS®	iHANS	
F_{zt} (kN)	1.9	0.9	1.5	1.7	3.2
F_{zc} (kN)	1.2	2	1.5	3	3.2
Max. N_{ij}	0.4	0.6	0.4	0.7	1.0

It will be noted that during the early phase of the event (i.e. before 80ms), the iHANS generates neck injury risk numbers that are (with the exception of tensile loading in the frontal test) lower than those of the standard setup. In fact Nij is markedly lower in both the frontal and offset tests. During the rebound phase of the event (Table 2) the standard setup fares better than iHANS. Though Nij for both tests are below the failure criterion, the 3kN axial compression with iHANS late in the offset test is close to the failure limit of 3.2kN. This particular observation requires further examination or testing.

Review of the high-speed videos showed that upon impact, the ATD head translates forward until captured by the helmet interior. The head and helmet then move as one until the crash pulse is over. In the frontal and the offset tests, the overall head motion is quite *airbag*-like. That is to say, the forward head excursion and the neck flexion are greatly limited. Figure 10 shows the position of the ATD at maximum forward excursion during a frontal sled test.

Figure 10: ATD with iHANS prototype at maximum forward excursion.

Direct Impact Tests

Currently, auto racing helmets are assessed with a direct impact protocol typified by that of the Snell Foundation Standard SA(2010) or the Federation Internationale d'le Automobile FIA8860(2004). In those tests, the helmet is mounted upon a metal headform and the assembly is dropped from a predetermined height onto a variety of steel anvils. The headform linear acceleration is monitored and the maximum acceleration and (in FIA8860) HIC are recorded.

Insofar as direct impact is still a threat to a driver's head, it was considered appropriate to conduct tests of this nature. However, dropping the *helmet* portion of the system onto a steel anvil was not appropriate because the system relies on support by the shoulders and restraint belts. Furthermore, it was considered important to monitor for rotational motion as well.

The protocol developed here entailed a stationary seated Hybrid3 ATD wearing either an experimental iHANS or wearing a new Bell auto racing helmet attached to a sliding tether HANS®. A test buck was fabricated that simulated the seating geometry described in SFI 38.1 and allowed the use of standard 5-pt racing harnesses. A full Hybrid3 pelvis and upper torso was seated in this buck. The Hybrid3 headform was instrumented with a nine accelerometer cluster to facilitate the calculation of angular head accelerations.[5] A six axis load cell was used to measure neck loads. Impacts were delivered via a linear impactor system.

Preliminary tests were conducted on a new standard setup and the same prototype that had been subject to the sled tests.[6] Other tests were conducted on new undamaged prototypes. Impacts ranged from 225J to 450J to the front, sides, rear and crown of the helmet. The impactor was either a steel flat plate or a steel 48mm radius hemisphere (in keeping with contemporary helmet standard test methods).

The results of this preliminary testing suggested that iHANS neck and linear head injury metrics were generally lower than the standard setup. However there were several issues that needed to be resolved. Among these was a higher maximum angular acceleration generally associated with iHANS compared to the standard setup.

From the high speed videos, it was observed that stretching of the belts overlying the collar occurred as the lateral head motion is restrained. This is an interesting additional load limiting (energy absorbing) feature not present with a standard

[5] They Hybrid3 head was originally intended for frontal impact, but has since been adopted for use in side impact crash testing as the SID-H3 model.

[6] Because of the viewing port in one of the iHANS prototypes, some test sites were not practical.

helmet/HRS setup. Selected frames from high speed video are shown in Figure 11 and Figure 12. It should be noted that the Hybid3 head and neck assembly is seldom used for side impact work (though specified in the SID H3 side impact dummy) as the lateral stiffness is not entirely biofidelic. Nevertheless, for current purposes of response comparisons, the consequence is felt to be minor.

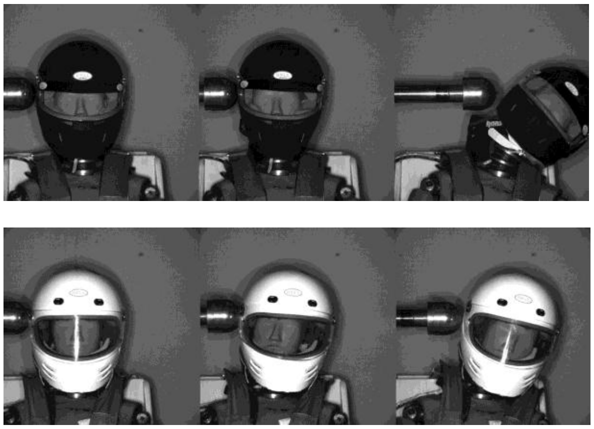

Figure 11: These images show the standard (top) and iHANS (bottom) responses at initial contact, maximum liner compression and maximum head excursion for the 225J side hemi impact.

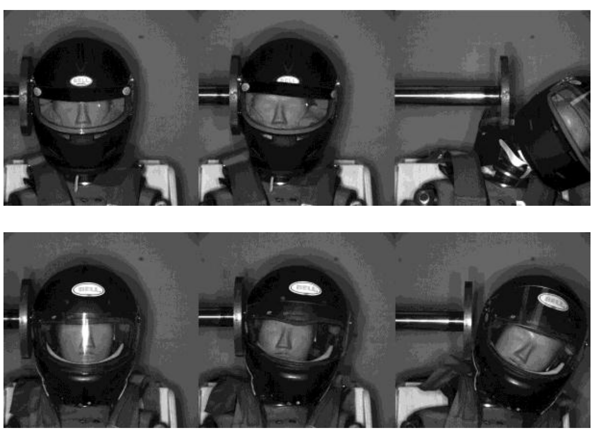

Figure 12: These images show the standard (top) and iHANS (bottom) responses at initial contact, maximum liner compression and maximum head excursion for the for the 450J flat side impact.

Following the above observations, an additional series of tests was conducted. The same basic protocol as the previous tests was employed. Flat steel surface impacts were delivered to the top, front, rear and side of the headgear. Steel hemispherical impacts were delivered to the side. However this time, both helmet shells and EPS liners were the same models (though of different sizes); the SA 2010 Bell model M4. A layer of low density open cell foam was installed throughout the interior liner of the iHANS helmet. This would serve to cushion the initial contact between the head and the liner. Another difference in this series was that for side hits, the seat buck was tilted 15 degrees so as to contact the helmet in its standard impact test zone. All impacts were delivered at nominally 450Joules.

A typical test configuration is shown in Figure 13.

Figure 13: Side impact configuration.

To analyze the test data further, the Simulated Injury Monitor (SIMon) finite element head model developed by NHTSA (Takhounts et al 2008) was exercised. This model allows one to compute brain distortion pattern histories but more importantly, generates several criterion functions that have been (more or less) correlated to the severity of various brain injury types in humans.[7] It will be understood of course that any contemporary FEM brain model is simply a tool whereby trends can be predicted and comparisons made. They have not yet developed to the point where absolute accurate results can be generated.

Takhounts et al (2011) recently published a new criterion function “The Kinematic Brain Injury Criterion BRIC”. This empirical function has been correlated quite well to CSDM. Its merit is that it completely circumvents the SIMon model and allows

[7] In particular;

- The cumulative strain damage measure CSDM for diffuse axonal injury DAI
- The relative motion damage measure RMDM for acute subdural hematoma ASDH and
- The dilatational damage measure DDM for contusion and focal lesions.

In addition, the model generates a parameter that is known to correlate to ASDH, the maximum principal strain MPS. The NHTSA authors have recently expressed reservations about the validity of the newest SIMon model for RMDM and for DDM so these indices are not reported here.

one to directly examine rotationally induced injury probability from the recorded Hybrid3 head kinematics. It evolved from simulations of frontal impact tests with the Hybrid3 (NCAP tests from NHSA database) and side impact tests with the ES2 test dummy and WorldSID. SIMon has not been validated for every conceivable impact scenario and BRIC no doubt will be refined as more human head injury tolerance data becomes available. In the meantime, these models do provide insight into the head injury probabilities in the present test series.

Summaries of the pertinent test results are shown below in Table 3. Neck responses are shown first followed by linear head responses then combined or rotational head responses. To illustrate the comparative response of HANS® with the prototype iHANS, these peak data are normalized by critical thresholds and presented in Figure 14 in percentages. Paired tests are side-by side. Note that many of these critical thresholds might be debated, however the objective is simply to illustrate several peak metrics against one common scale.

There are several observations of note:

- For side impacts, the neck injury risk is markedly less with iHANS than with the standard setup.
- Rear impact responses to the prototype and to the standard setup are about the same for both head and neck injury.
- For crown impacts the iHANS results in about one quarter the neck compressive loading with a commensurate lower Nij.
- For frontal impact the standard setup generally yielded lower neck injury risk. The neck shear force and bending moments with the standard setup were lower than the iHANS. However, the neck tensile loading was twice that of the iHANS.
- Head injury due to crown loading is no more likely with one system or the other.
- For the combined/rotational head responses, the maximum principal strain computed with SIMon appears largely independent of system type. It is lowest for frontal impacts.
- For frontal and lateral impacts, virtually all head injury metrics are lower for iHANS than for the standard setup.
- In spite of somewhat higher angular acceleration for iHANS, those metrics that incorporate angular acceleration and angular velocity (which is generally lower with the iHANS prototype) indicate that the iHANS produces lower risk of rotationally induced brain injury in all but the crown loading (which is low to begin with) under these test conditions.

Table 3: Neck and head responses from 450J hemispherical and flat direct impacts.

450 J		side R48mm hemi		side flat		front flat		rear flat		crown flat		critical threshold
		Helmet w/HANS®	iHANS	Helmet w/HANS®	iHANS	Helmet w/HANS®	iHANS	Helmet w/HANS®	iHANS	Helmet w/HANS®	iHANS	
neck (maxima)	M_x (Nm)	42.3	25.8	43.3	28.4	*	*	*	*	*	*	n/a
	M_y (Nm)	*	*	*	*	66.5	102.9	47.1	43.1	180	61.3	n/a
	F_x (N)	*	*	*	*	1086	1482	1214	1227	*	*	n/a
	F_y (N)	1119	769	1164	992	*	*	*	*	*	*	n/a
	F_{zt} (N)	2215	1181	1828	1060	2403	1135	1420	464	602	829	4000
	F_{zc} (N)	2395	1010	3512	1600	756	486	770	834	11348	2812	4170
	Max. N_{ij}	0.47	0.26	0.66	0.37	0.85	0.93	0.56	0.49	3.18	0.91	1.0
head	A_{max} (G)	183	146	175	162	158	116	173	196	135	82	300
	HIC	898	653	975	834	625	375	1023	926	361	141	1000
	α_{max} (rad/s^2)	12437	13060	13209	16002	8077	5644	9691	9575	3639	5451	10000
	ω_{max} (rad/s)	45.7	36.0	45.0	33.5	43.9	35.5	42.5	38.3	10.1	15.7	n/a
	HIP (kW)	28.4	25.8	31.1	29.9	36.2	23.5	36.4	33.2	13.9	10.4	30
	BRIC	1.30	1.10	1.30	1.12	1.15	0.91	1.16	1.07	0.31	0.48	1.30
	MPS	0.54	0.57	0.54	0.56	0.11	0.09	0.54	0.56	0.08	0.08	0.4
	CSDM (0.25)**	0.13	0.11	0.13	0.09	0.00	0.00	0.00	0.00	0.00	0.00	0.56

* secondary importance due to loading direction.
** volume fraction of the total brain exceeding a strain of 0.25
n/a denotes not applicable

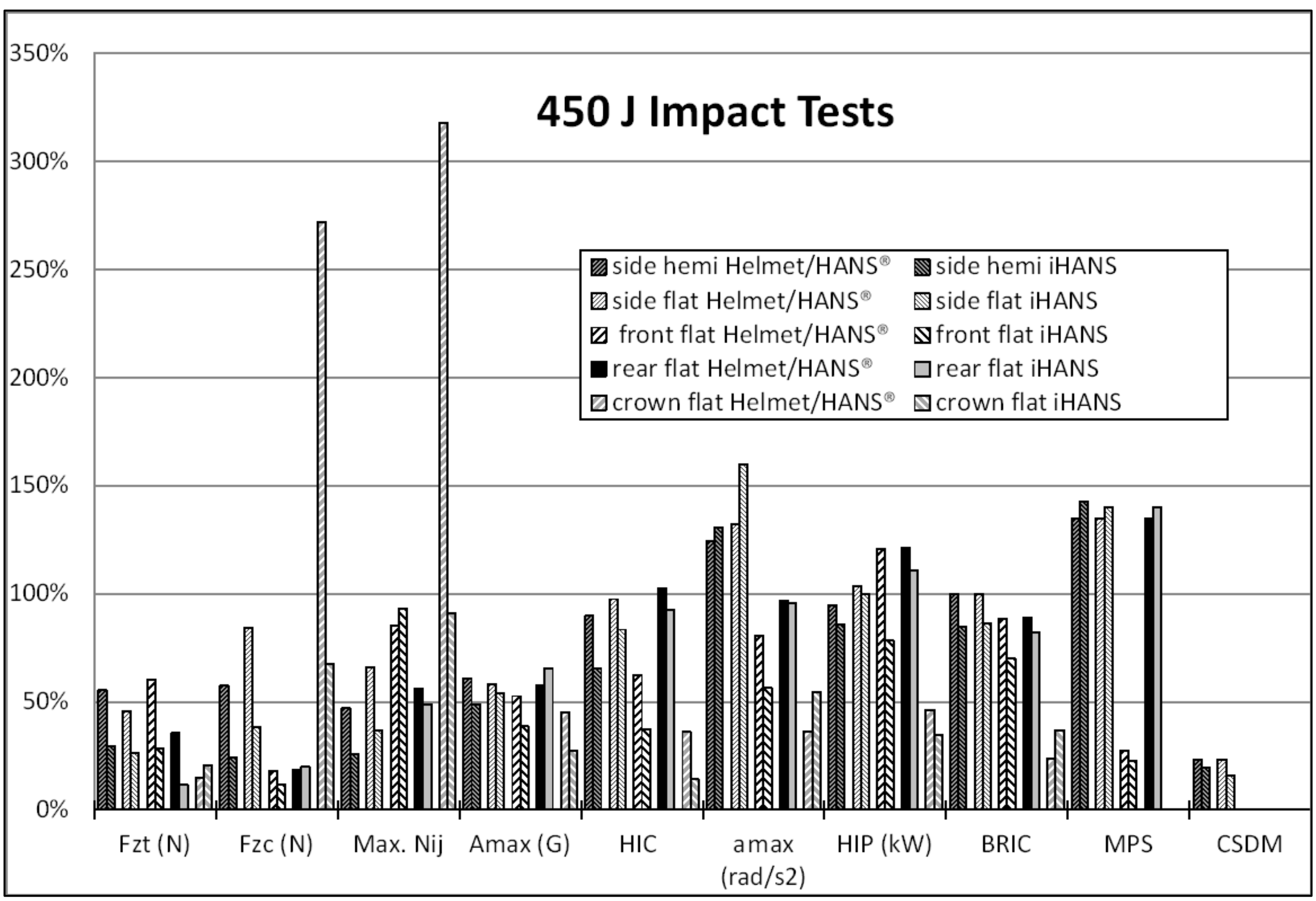

Figure 14: 450J direct impact normalized neck and head response metrics

Ballistic Testing

Resistance to penetration by a helmet is predicated partly on the fact that the customary helmet is in direct contact with the head. This provides a support for the helmet assembly and assists in providing load distribution. The iHANS, however, will stand off from the head and as such is essentially a domelike structure that can deflect and/or collapse at the location of the impact possibly leading to concentrated loading of the head. On the other hand, much of the energy of the impact is dissipated in the process so the probability of localized loading may in fact be reduced compared to a standard helmet. These tests were intended to explore this issue.

As previously, the units were mounted on a Hybrid3 torso with the head and neck instrumented with the usual acceleration cluster and a six axes neck load cell.

Using a pneumatic cannon, various size steel balls were fired at various sites on the standard and experimental iHANS at velocities up to 44m/s. Though significant shell damage occurred to both systems, neck and head loading were minimal for both the iHANS prototype and the standard setup.

Following these preliminary tests, a new test device was utilized which can be seen in Figure 15. This headform was developed principally to assess blunt impact threats associated with ballistic helmet loading or other direct impact scenarios (Anctil et al, 2005). It is modeled on an ISO headform shape but is fitted with an array of load sensors at the forehead region. The headform does not have the biofidelic properties of the Hybrid3 but for these kinds of tests, that is not entirely necessary. However, like the Hybrid3, the forehead (and the transducer array) is covered with a layer of molded rubber fitted to complete the ISO forehead geometry.

Figure 15: Load sensing headform.

The air cannon was positioned as shown in Figure 16. A 51mm diameter steel ball was fired at 44m/s (100mph) at the center of the forehead region of the helmet overlying the load sensing array. The events were captured on high speed video.

In both cases the exterior shell was badly damaged but, in both cases, the projectile was deflected away from the head and complete penetration of the shell did not occur. The maximum force measured in both cases was marginally less than that suggested as a threshold for skull fracture (Viano et al, 2004) and the two systems differed by only 50N.

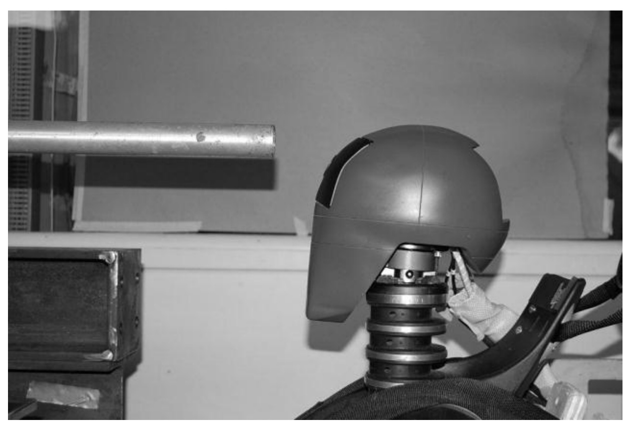

Figure 16: Ballistic cannon test setup. (Note the metal bar indicates the trajectory of the steel ball. Head protection was added afterwards.)

DISCUSSION AND CONCLUSIONS

Results of the on- track trials with iHANS prototypes suggest that the new head/neck protection system can mitigate most of the undesirable attributes of the customary helmet. The reintroduction of soft interior comfort padding should ease concerns about the head interacting with the helmet interior during cornering. However, far more extensive, long-term in-car driver track testing will be needed to refine driver operational features such as ease of donning/doffing, wind noise, adaptation to different size heads and necks and to different driver seating postures.

The limited sled testing to date suggests that the formal requirements for head and neck restraint can be achieved and that the integrated helmet and neck support system is as good as or better than a contemporary system in limiting neck loading.[8] However, the standard test cannot by itself quantify the overall performance of the system. It will be important to establish that the SFI 38.1 specifications can be met with a variety of seat angles. Indeed, sled tests should be run with the more biofidelic THOR thorax and shoulder structure. A more advanced dummy neck design, when developed, should also be included in future sled testing. As important perhaps, will be to establish sled-based performance under pure lateral loading using appropriate side impact ATDs. Perhaps, in the future, sled testing with post mortem human subjects might be considered to possibly uncover unexpected injuries and loading mechanisms.

The direct impact test program was based largely on contemporary helmet performance standards. Customary impact sites and impact anvils were used. However, by the nature of the experimental system, a different protocol was used. By adopting the Hybrid3 head and neck, not only were linear head accelerations measured but so too were rotational accelerations as well as neck reaction loads. Comparison of bending moments, axial forces and Nij between the standard and experimental setups showed that iHANS was generally as good as or better at reducing neck injury risks under these test conditions. By virtue of its design, transferring load down to the shoulders, it provides significant reduction of axial compressive loading of the neck when the helmet portion is loaded. Similarly a comparison of maximum linear acceleration, HIC, HIP, BRIC, MPS and CSDM between the standard and experimental setups showed iHANS to be generally as good as or better than the standard setup at reducing brain injury risks under these test conditions. Again, the absolute reduction in injury risks is unknown but using these injury assessment functions, iHANS appears to have promise. Further testing at different sites and at different energy levels will be necessary to optimize the shell and interior energy management system while ensuring adequate head space during normal operation.

It is of considerable interest that, in these tests, the presence of a gap between the helmet interior and the headform is not deleterious (indeed positive) to head response. The manner by which the head interacts with the helmet cannot be discerned in detail but it is likely that the influence is partly the same as that of a thick very soft comfort liner. The energy absorbing EPSB liner in a customary helmet cannot function until the much softer comfort liner is compressed and it can then bear directly on the wearer's head. The speed with which the liner makes contact with the head is partly governed by the inertia of the helmet itself as it accelerates toward the head. With the greater standoff of the iHANS, its additional mass and the fact that its movement is restrained by the collar/belt attachment, the speed of liner/head interaction would likely be significantly less. In an extreme case (with lower impact energy or greater standoff), the iHANS liner would not make contact with the head at all and the head reaction load and acceleration would be zero. With a standard close fitting helmet there would always be some reaction load on and acceleration of the head.

The manner by which lateral head motion is restrained by the iHANS in side impacts is of particular interest in that it appears to help reduce the probability of rotationally induced brain injury as well as reducing the chances of lateral head impact in vehicles not equipped with head lateral restraint.

Finite element modeling of the brain, which has been employed here as a tool, holds promise for better predicting the detailed nature of brain distortion under complex loading such as this. Whether they have achieved an adequate level of sophistication to do this today is questionable.

The ballistic testing was limited to a steel ball striking the helmeted head at 100mph. On the basis of load maxima and distribution on the dummy forehead, the experimental prototype does not appear to be at a disadvantage regarding skull fracture risk compared to an identical contemporary helmet. In fact the opportunity exists to strengthen the shell substantially with the iHANS and not suffer the head weight-borne penalty of the standard setup.

The overall objective of this work has been to get the helmet (as we know it) off the race car driver's head

[8] Insofar as neck injury assessment is concerned, it will be appreciated that Nij may not provide all the confidence one might wish for as it does not include direct or torsional shear in its formulation.

while maintaining the same or lower injury risk that a helmet and a modern HRS provides.

Notwithstanding the progress made, this study has admittedly been no more than examining the biomechanical feasibility of a new concept. There is a myriad of issues to be resolved before this prototype can be transformed into a practical device. Such practical considerations would be the responsibility of potential manufacturers now that operational and biomechanical feasibility has been verified.

Agencies concerned with race car driver safety are encouraged to conduct their own tests of this new head and neck restraint system.

ACKNOWLEDGMENTS

Many individuals and organizations contributed in many ways to this project. Special thanks to

- John Tutton, Luc Boucher and Romeo Graham of DW Product Development in Ottawa, Canada who generated the original concept drawings and produced the first wearable mockups.
- Kyle Kietzmann and Ed Rothrock of Bell Auto Racing and Jim Downing of HANS® Performance Products for assisting with prototype development.
- Nigel Mortimer, David Murray and Jim Downing, all expert race car drivers who trusted us to protect them during numerous on-track trials, and whose feedback was truly valuable.
- The staff at MGA who graciously accommodated our somewhat unorthodox testing requirements and schedule.
- John Melvin of Tandelta Inc., Hubert Gramling of FIA and Bob Hubbard, Professor Emeritus of Michigan State University. Each of them have provided important technical input and have *kept us realistic* in terms of what are reasonable expectations for this kind of development.
- Liying Zhang of Wayne State University who helped us deal with some of the mathematical brain modeling issues.

REFERENCES

Anctil, B., Keown, M., Bourget, D., and Pageau, G. (2005) A Novel Test Methodology to Assess the Performance of Ballistic Helmets. 22nd International Symposium on Ballistics, Vancouver, Canada.

Federation Internationale de l'Automobile. FIA 8860-2004 Advanced Helmet Test Specification.

Foster, J.K., Kortge, J.O. and Wolanin, M.J. (1977) Hybrid III – A Biomechanically-Based Crash Test Dummy. Proc. 21st Stapp car Crash Conference, pp 973-1014. Society of Automotive Engineers, Warrendale PA.

Gohmert, D. M. (2012) Personal communication re helmets for Orion crew members.

Gramling, H. (2012) Personal communication regarding FIA sponsored ATD neck development.

Grohs, T. M. and Archer, S. (2000) Head Impact Protection Developments in the F1 Cockpit with Investigations to Correlate Head Movement to Brain Injury. SAE Technical paper 2000-01-3558, presented at the Motorsports Engineering Conference and Exposition, Dearborn, MI.

Haffner, M., Eppinger, R., Rangarajan, N., Shams, T., Artis, M. and Beach, D. (2001) Foundations and Elements of the NHTSA THOR Alpha ATD Design. Presented at the 17th ESV Conference, Amsterdam.

Hubbard, R. P. (1987) Neck Protection Device with Occupant of a High Performance Vehicle. US Patent No. 4568510.

King, A. I., Yang, K. H., Zhang and Hardy, W. (2003) Is Head Injury Caused by Linear or Angular Acceleration? Proc. International IRCOBI Conference, Lisbon.

Melvin, J.W. (1988) The Engineering Design, Development, Testing and Evaluation of an Advanced Anthropomorphic Test Device, Phase 1: Concept Definition. US Department of Transportation, DoT-HS-807-224.

Melvin, J. W., Baron, K.J., Little, W.C., Gideon, T.W., Pierce, J., (1998) Biomechanical Analysis of Indy Car Crashes. Proc. 42nd Stapp Car Crash Conference, pp 247-266. Society of Automotive Engineers, Warrendale PA.

Melvin, J.W. (2012) Personal communication re development of SFI 38.1 standard.

Melvin, J. W. and Gideon, T., (2004) Biomechanical Principles of Racecar Seat Design for Side Impact Protection. SAE Technical paper 2004-01-3515, presented at the Motorsports Engineering Conference and Exposition, Dearborn, MI.

Mertz. H., Prasad, P., and Nusholtz, G. (1996) Head Injury Risk Assessment for Forehead Impacts.

SAE Technical paper 960099, Society of Automotive Engineers, Warrendale PA.

Newman, J.A., Shewchenko, N., Welbourne, E., (2000) A Proposed New Biomechanical Head Injury Assessment Function – The Maximum Power Index. Stapp Car Crash Journal 44: 215-248.

SFI Foundation Inc. Specification 38.1. (2011) Head and Neck Restraint Systems.

Takhounts, E. G., Ridella, S. A., Hasija, V., Tannous, R. T., Quinn Campbell, J., Malone, D., Danelson, K., Stitzel, J., Rowson, S., and Duma, S. (2008) Investigation of Traumatic Brian Injuries Using the Next Generation of Simulated Injury Monitor (SIMon) Finite Element Head Model. Stapp Car Crash Journal 52: 1-31.

Takhounts, E. G., Hasija, V., Ridella, S. A., Rowson, S., and Duma, S. M. (2011) Kinematic Rotational Brain Injury Criterion (BRIC). Proceedings of the ESV Conference, Washington.

Viano, D. C., Bir, C., Walilko, T. and Sherman, D. (2004) Ballistic Impact to the Forehead, Zygoma and Mandible: Comparison of Human and Frangible Dummy Face Biomechanics. Journal of Trauma-Injury, Infection and Critical Care 56(6): 1305-1311.

APPENDIX:

INJURY ASSESSMENT FUNCTIONS

Various head and neck injury assessment functions are employed in this study and are summarized here. All are based upon measurements of a mechanical surrogate (e.g. Hybrid3 ATD) or predictions of mathematical surrogates (e.g. SIMon).

Head

A_{max} = The maximum value of the resultant linear acceleration of the center of gravity of the test headform. Usually expressed in gravitational units, it is used extensively in the evaluation of helmet impact performance.

α_{max} rad/s/s = The maximum value of the resultant angular acceleration of the test headform. Not, by itself, considered a good correlate to brain injury and not specified in any published performance standard.

HIC sec. = NHTSA's Head Injury Criterion.

$$\mathrm{HIC} = \left[\frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} a(t)dt \right]^{2.5} (t_2 - t_1)$$

First referenced in 1974, it is the current assessment function for the evaluation of closed head injury probability in automotive frontal crash testing and certification. It relies solely on a portion of the resultant linear acceleration history of the ATD head following impact.

HIP kW = Head Impact Power.

$$\begin{aligned} HIP = {} & 4.50\, a_x \int a_x dt + 4.50\, a_y \int a_y dt \\ & + 4.50\, a_z \int a_z dt + 0.016\, \alpha_x \int \alpha_x dt \\ & + 0.024\, \alpha_y \int \alpha_y dt + 0.022\, \alpha_z \int \alpha_z dt \end{aligned}$$

Developed by Newman et, al (2000), this function combines linear and angular acceleration and velocity into a single closed head injury assessment function.

MPS = Maximum Principal Strain from SIMon (Takhounts et al 2008)

CSDM = Cumulative Strain Damage Measure; i.e. the volume of the brain for which the principal strain exceeds 0.25 from SIMon.

BRIC = Kinematic Brain Injury Criterion (Takhounts et al 2011), endeavors to circumvent the actual use of SIMon with an empirical relationship between the cumulative strain damage measure CSDM and rotational kinematics of the head.

ω_{max} and ω_{cr} are maximum and critical rotational velocities respectively and,

α_{max} and α_{cr} are maximum and critical rotational accelerations respectively. The critical values are different for the various dummies. Those for the Hybrid3 were used in the present analysis.

Neck

FMVSS-208 Neck Injury Criterion Nij is an empirical function that combines the relative contributions of axial loading and bending moments to assess neck injury probability.

F_z = maximum measured axial load;

F_{int} = critical intercept value;

M_y = maximum measured moment at the occipital condyle;

M_{int} = flexion/extension moment critical intercept value.

- When calculating Nij the critical intercept values employed were:
- *Fzc* = 6806N when *Fz* is in tension
- *Fzc* = 6160N when *Fz* is in compression
- *Myc* = 310Nm when a flexion moment exists at the occipital condyle
- *Myc* = 135Nm when an extension moment exists at the occipital condyle.
- For side impacts a critical lateral extension bending moment of 135Nm was used.